other than drop foot. Both participants had an absence of strongly manifesting spasms and contractures in lower extremity joints. Finally, each participant had used an AFO for at least two years and therefore was experienced at AFO ambulation. Subjects reached a stable neurological state after the incident that caused their disability. Thus, no recovery of function was expected or found. Three normal subjects were matched for gender, height, weight, and age to the drop foot participants. Subject sex, age, mass, height, and self-selected gait speed are listed in Table III.

TABLE III

Subject	Sex	Age (yr)	Mass (kg)	Height (m)	Self-Selected Gait Speed (m/s)
Drop Foot	M	62	79.1	1.79	1.22
Drop Foot	M	62	95.4	1.77	1.07
Normal	M	66	76.6	1.70	1.39
Normal	M	67	86.1	1.75	1.01
Normal	M	67	73.2	1.70	1.22

[0040] Kinematic and kinetic data were measured on both the affected and unaffected sides using an eight-camera VICON **512** system and two AMTI force plates. The data were processed at 120 Hz with VICON Workstation using the standard model of the lower limbs included with the software. These data were then analyzed using MATLAB.

[0041] The subjects donned the AAFO in three different control conditions: zero, constant, and variable impedance. The zero impedance control setup was implemented by setting the target force on the SEA to zero, thereby minimizing the impedance contribution of the orthosis across the ankle joint. This setup was meant to approximate unassisted drop foot gait. For the constant impedance control setup, the AAFO controller commanded a constant joint stiffness, independent of walking phase and gait speed. This joint stiffness was the converged controlled plantar flexion (CP) stiffness from the variable impedance control that minimized the number of slap foot occurrences at the self-selected gait speed. This constant impedance control condition was designed to imitate conventional AFO technology employed in the treatment of drop foot gait.

[0042] For each controller, subjects walked at slow, self-selected, and fast gait speeds. The subjects first walked at their self-selected speed using the constant impedance control setup. The amount of time required to cover a specified distance was measured using a stopwatch. Subjects were then asked to reduce their time by 25% for the fast gait speed and increase their time by 25% for the slow gait speed. These times were then matched when testing the remaining two control conditions.

[0043] A stride cycle was defined as the period of time for two steps, and was measured from the initial heel contact of one foot to the next initial heel contact of the same foot. All data were time normalized to 100% of the stride cycle. The ankle angle data during a gait cycle were fitted with a cubic spline function and then resampled to 200 samples so that each point was 0.5% of the gait cycle.

[0044] In this study, it was assumed that normal gait was symmetrical and that deviations from a reference pattern were a sign of disability. To analyze spatial asymmetry, the step length on the affected side ($L_{affected}$) was subtracted from the step length on the unaffected side ($L_{unaffected}$) The difference in stride lengths (L_{sym}) should be zero for symmetric gait:

$$L_{sym}=L_{affected}-L_{unaffected} \quad (1)$$

[0045] To analyze temporal asymmetry, the step time on the affected side ($T_{affected}$) was subtracted from the step time on the unaffected side ($T_{unaffected}$). The difference in stride times (T_{sym}) should be zero for symmetric gait:

$$T_{sym}=T_{affected}-T_{unaffected} \quad (2)$$

[0046] A multiple comparison using a one-way analysis of variance (ANOVA) was used to determine which means were significantly different for the gait symmetry. P values less than 0.05 were considered significant for all tests.

[0047] The first evaluation of the drop foot controller was to test whether the system was capable of converging to a final CP stiffness that reduced or prevented slap foot. For each of the three gait speeds, the controller was able to converge to a final stiffness value within 32 steps (**FIG. 5**). The CP stiffness increases with increasing gait speed. During the stiffness convergence at each of the three gait speeds, the occurrences of the high frequency forefoot force signal (typical of slap foot; see **FIG. 4A**) were reduced.

[0048] As a measure of the slap foot complication, the average number of occurrences of slap foot per 5 steps (25 steps total) were calculated for each drop foot subject, control condition, and gait speed (n=5). The participants were unable to walk at the fast gait speed using the zero force condition because it was not deemed safe. The constant impedance condition eliminated the occurrences of slap foot at the slow and self-selected gait speeds (**FIG. 6**). The three curves correspond to zero, constant, and variable impedance control conditions. However, slap foot occurrences increased at the fast gait speed. By adjusting CP stiffness with gait speed in the variable-impedance control condition, the number of occurrences of slap foot was reduced at the fast gait speed by 67% compared to the constant stiffness condition.

[0049] To quantify the reduction of the second major complication of drop foot, or toe drag, the swing dorsiflexion angular range was used. The dorsiflexion angular range was defined as the maximum plantar flexion angle during the powered plantar flexion phase of stance minus the maximum dorsiflexion angle during swing. The variable impedance control was able to increase the amount of swing dorsiflexion as compared to the constant impedance condition by 200%, 37%, and 108% for slow, self-selected, and fast gait speeds, respectively (**FIG. 7**). All data points for the normal participants are an average of 15 trials, whereas for the drop foot participants the averages are over 20 trials.

[0050] A constant impedance ankle-foot orthosis hinders powered plantar flexion (PP) since a dorsiflexion moment will be exerted against the foot during late stance. As expected, the constant impedance condition reduced the PP angle as compared to the zero impedance condition and the normals (**FIG. 8**). Here the PP angle was defined as the maximum plantar flexion angle during power plantar flexion minus the maximum dorsiflexion angle during controlled dorsiflexion in stance. The variable-impedance controller had a larger PP angle than the constant impedance control condition by 89%, 25%, and 82% for the slow, self-selected, and fast gait speeds, respectively.